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Wrist motion in handrim wheelchair propulsion

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Abstract—Prevalence rates of carpal tunnel syndrome (CTS) in the wheelchair user population are high. One of the possible causes of CTS in this population is the movement pattern of the wrist during handrim wheelchair propulsion, which could include large wrist joint angles and wrist/finger flexor activity. Combined with the repetitive character of the movement, this could, in time, be detrimental to the soft tissue of the wrist.

To study peak wrist joint angles and their relationship with wrist- and finger-flexor activity, a three-dimensional (3-D) analysis of wrist movement during the push phase was performed. Nine subjects (five nonimpaired controls, four wheelchair users) propelled a handrim wheelchair on a treadmill at three different velocities (0.83, 1.11, and 1.39 m/s) and three slopes (1, 2, and 3%), while the surface EMGs of the wrist- and finger-flexor group were recorded.

Average peak wrist joint angles during the push phase were: ulnar deviation, $-24 \pm 11^\circ$; radial deviation, $13 \pm 12^\circ$; flexion, $-14 \pm 18^\circ$; and extension, $34 \pm 16^\circ$. The values for ulnar and radial deviation were close to normal values for maximal range of motion (ROM) found in the literature. Peak extension was approximately 50% of ROM. The peak angles, which occurred with concurrent activity of the wrist flexors, were: ulnar deviation, $-22 \pm 11^\circ$; radial deviation, $13 \pm 10^\circ$; flexion, $-16 \pm 15^\circ$; and extension, $32 \pm 16^\circ$. The large deviation and extension angles, especially those recorded simultaneously with wrist flexor activity, are serious risk factors for CTS. This finding may help explain the high rates of CTS in the wheelchair user population.

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Key words: *biomechanics, carpal tunnel syndrome, handrim wheelchair propulsion, kinematics, 3-D motion analysis, wrist.*

INTRODUCTION

In the wheelchair user (WU) population, all activities of daily living, including locomotion and weight-bearing, are performed by the upper limbs. It is thus not surprising that overload injuries of the upper limb are a common problem (1). One of the most vulnerable areas is the wrist, which is often affected by carpal tunnel syndrome (CTS), with symptoms related to compression of the N. Medianus (2,3). Prevalence rates for CTS for individuals with spinal cord injury (SCI) may be as high as 50 to 60 percent and increase with the duration of the injury (4,5). It has been suggested that repetitive extreme excursions of the wrist, as well as direct external force, are the major risk factors in development of CTS (6), while concurrent activity of the wrist flexor muscles could be an additional risk (7). Some examples of potentially harmful activities are transfers and the regular elevation from the seat by locking the elbow and wrist (4), but the most important one may be the use of a handrim wheelchair for ambulation. This might be due to the necessity of gripping and holding the handrim, which could force the wrist into extreme excursions. Also, the high frequency of movements might have a detrimental effect. Burnham et al. (8) found acute changes in nerve conduction velocity, related to CTS, elicited by handrim wheelchair propulsion. In addition, Gellman et al. (4)

found a serious elevation of pressure in the carpal tunnel when the wrist was in full active flexion or full active extension. They suggest that the etiology of CTS in the WU population may be a combination of the repetitive trauma from the propulsion of a handrim wheelchair and of ischemia resulting from increases in pressure in the carpal tunnel during extreme extension. Combination of the above indicates that wrist movement and muscular activity pattern during handrim wheelchair propulsion may be important factors in the development of CTS in the WU population.

The suggestions by Gellman et al. (4) are supported by overviews of occupational risk factors for CTS. Armstrong et al. (9) mention repetitive work with the hand, work that involves repeated wrist flexion or extreme extension, and repeated forces on the base of the palm and wrist.

While extreme wrist joint angles during wheelchair propulsion thus appear to be an important risk factor for CTS, they have not been studied extensively. To the knowledge of the authors, three studies have been published that describe wrist angles (10–12). This is most likely due to the three-dimensional (3-D) nature of the motion. Results of those studies have indicated that wrist angles during wheelchair propulsion could exceed levels reported as hazardous in the ergonomic literature (13) and could become close to values published on the active range of motion (ROM; 14–17).

The aim of this study was to determine whether wrist motion during wheelchair propulsion reaches the active ROM. It is hypothesized that flexion-extension and radial-ulnar deviation angles approach the ROM during the push phase of wheelchair propulsion, and that these limits are even approached when the wrist flexors are active.

METHODS

Nine male subjects (age 25–40 years, weight 65–113 kg) participated in this study. Four used a wheelchair on a regular basis due to SCI (lesion level below T4, time since injury 5–14 years); five were nonimpaired controls, some of whom had prior experience in handrim wheelchair propulsion. All gave written informed consent and were in good health, without current upper limb disorders.

The experiment was conducted on a motor-driven treadmill (Enraf Nonius 3446, Delft). Eight subjects drove a modified block-frame basketball wheelchair

(Morriën Tornado, handrim diameter 0.54 m). One subject had to use his own wheelchair (handrim diameter 0.51 m). The experiment was comprised of two tests, separated by at least a 10-min rest and preceded by a familiarization period. Both tests consisted of three 3-min exercise bouts. The first was on a constant slope (2 percent) at three successive velocities (0.83, 1.11, and 1.39 m·s⁻¹ or 3, 4, and 5 km·h⁻¹). In the second, the slope was increased after each third minute (1, 2, and 3 percent) at a constant velocity (1.11 m·s⁻¹). These conditions resulted in a mean required power output of 26, 34, and 43 W for the first test and 22, 34, and 48 W for the second. The combination of conditions allowed for the evaluation of the effect of speed and slope on the kinematics of the wrist.

During the tests, data were collected in each third minute at the right side of the body on at least three consecutive propulsion cycles.

Motion data were collected with an opto-electronic 3-D motion-analysis system (Vicon, Oxford Metrics, Oxford, UK; cameras: MOS-tv, 60 Hz). Reference markers were placed on the wheelchair axis and frame. The arm, forearm, and wrist were equipped with cuffs with four markers each, placed on the dorsolateral side of the upper arm, on the dorsal side of the forearm as distally as possible, and at the dorsal side of the hand over metacarpals II and III. Since the forearm cuff was fastened on the most distal part of the forearm, it was assumed that its movements would closely follow pro-supination movements, as well as 'normal' elbow flexion. The cuffs on upper arm and forearm were attached by Velcro straps. The hand cuff was attached with an elastic bandage around the wrist and the distal part of the metacarpals (**Figure 1**).

A separate reference measurement was conducted to determine the relationship between the anatomical local coordinate systems and the cuff markers. This measurement was made with the arm in the anatomical position and additional markers on significant anatomical landmarks. **Table 1** summarizes the definitions of local coordinate systems and the anatomical landmarks that were used. The forearm coordinate system is defined on the basis of the medial epicondyle (EM), and both processii styloidei on the wrist. As a consequence, the forearm coordinate system will rotate with pro- or supination movements of the forearm.

From the positions of the cuff markers and anatomical landmarks, the relationship between cuffs and local coordinate systems could be determined:

Table 1.

Description of definitions of local coordinate systems for forearm and hand. These definitions only apply in the anatomical position.

Segment	Axis	Definition	Relevant landmarks
Forearm (Radius)	Y-axis	pointing from EM to PU	EM=Epicondylus Medialis
	Z-axis	perpendicular to PU - PR - EM	PU=Processus Styloideus Ulnae
	X-axis	perpendicular to Y and Z	PR=Processus Styloideus Radii
Hand	Y-axis	pointing from the midpoint between PU and PR to DM	PU=Processus Styloideus Ulnae PR=Processus Styloideus Radii
	Z-axis	perpendicular to PU - PR - DM	DM=Proximal end of Digitus Medius
	X-axis	perpendicular to X and Z	

$$\begin{aligned} CF_a &= CF_{ref} * RF_{ref}^T \\ CH_a &= CH_{ref} * RH_{ref}^T \end{aligned} \quad [1]$$

where:

CF_a, CH_a = the orientation of the cuff markers on forearm and hand in the anatomical position;

CF_{ref}, CH_{ref} = the orientation of the cuff markers during the reference measurement;

RF_{ref}, RH_{ref} = the orientation matrices of the local coordinate systems during the reference measurement.

Cuff data were arranged as row vectors.

Wrist rotations during the tests were calculated relative to the anatomical position.

The orientations of the forearm and hand in any recorded position $RF_{a,i}$ and $RH_{a,i}$ were calculated as the rotations of the cuffs relative to the anatomical position, using the least-squares algorithm described by Söderkvist and Wedin (18):

$$\begin{aligned} CF_i &= CF_a * RF_{a \rightarrow i} + \underline{f} \\ CH_i &= CH_a * RH_{a \rightarrow i} + \underline{h} \end{aligned} \quad [2]$$

where:

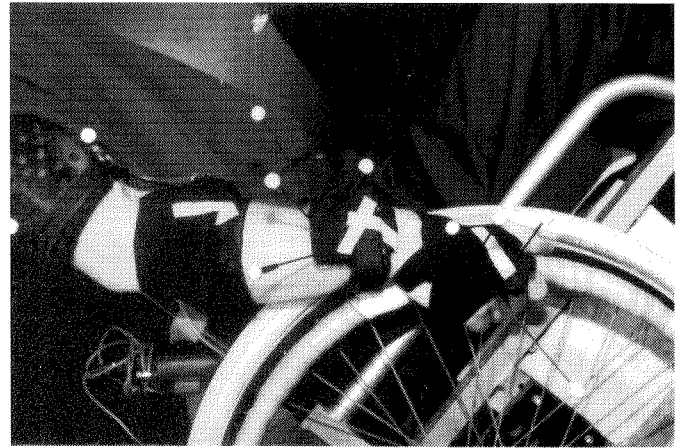
CF_i, CH_i = the orientation of the cuff markers during the tests;

$\underline{f}, \underline{h}$ = translation vectors (which are not used in this study);

$RF_{a \rightarrow i}, RH_{a \rightarrow i}$ describe the rotation from anatomical position to test position.

The rotations in the wrist (RW_i) were calculated as the difference between forearm and hand rotations:

$$RW_i = RH_{a \rightarrow i} * RF_{a \rightarrow i}^T \quad [3]$$

**Figure 1.**

Hand with cuff. Also given are the anatomical landmarks DM, PR, and PU. EM is in this view invisible. The co-ordinate systems are the local systems as defined in **Table 1**.

The joint rotation matrix RW_i was subsequently decomposed in Euler-angles. The wrist rotation matrix was decomposed with the sequence x-z'-y'':

$$RW_i = Rx(\gamma_{w,i}) * Rz(\alpha_{w,i}) * R^w y(\beta_{w,i}) \quad [4]$$

Wrist flexion and deviation were defined as the primary rotation angles around x-axis ($\alpha_{w,i}$) and z-axis ($\gamma_{w,i}$), respectively.

Hand length was measured to reconstruct the position of the third metacarpal (MCIII) between the proximal end of the middle finger (DP) and the line between the ulnar styloid (PU) and radial styloid (PR).

Table 2.

Peak values for wrist flexion and deviation angles, given for the six conditions and two subject groups. () indicate standard deviations.

	Control Subjects (N=5)				Wheelchair Users (N=4)			
	Ulnar deviation	Radial deviation	Extension	Flexion	Ulnar deviation	Radial deviation	Extension	Flexion
4 km/hr	-23	19	34	-3	-22	14	32	-26
1%	(10)	(9)	(27)	(18)	(14)	(13)	(11)	(10)
4 km/hr	-26	16	37	-4	-24	13	35	-31
2%	(8)	(8)	(18)	(14)	(16)	(13)	(9)	(10)
4 km/hr	-23	14	37	-1	-24	13	33	-28
3%	(7)	(9)	(20)	(16)	(13)	(11)	(12)	(10)
3 km/hr	-24	17	38	-1	-22	15	30	-29
2%	(11)	(8)	(18)	(14)	(13)	(14)	(12)	(5)
4 km/hr	-24	21	42	1	-26	7	28	-29
2%	(8)	(6)	(16)	(18)	(12)	(11)	(16)	(8)
5 km/hr	-20	3	29	-5	-26	9	24	-28
2%	(14)	(20)	(20)	(13)	(14)	(12)	(16)	(9)
Mean	-23 ^c	15	36 ^b	-3 ^a	-24 ^c	12	30 ^b	-28 ^a
(sd)	(9)	(12)	(19)	(14)	(12)	(11)	(12)	(8)

^a=significant difference between WU and controls; ^b=significant effect of speed; ^c=significant effect of speed and slope.

The joint angles were low-pass filtered (2nd order recursive Butterworth filter, $F_c=6$ Hz). Peak joint angles were determined for the push phase of each recorded cycle. The values for three consecutive cycles of each condition were averaged. In addition, kinematic data were combined with EMG data to obtain peak angles achieved during wrist flexor activity (WFA).

Surface EMG was recorded of the wrist flexor group with a pair of Ag/AgCl electrodes. The signals were preamplified (1,000 times) and transmitted telemetrically (Biomes 80, Glonner Electronics GmbH, Munich, Germany) to the VICON system. Linear envelopes were constructed by rectifying and low-pass filtering (4th order recursive Butterworth filter, $F_c=10$ Hz) of EMG signals, followed by resampling with a frequency of 60 Hz.

Electrodes were positioned at a mutual distance of 2 cm, along the longitudinal axis of the forearm and approximately 7 cm distal to the EM. To ensure correct placement, EMG values were evaluated during maximal finger and wrist flexion and maximal extension. During wrist flexion the maximal voluntary contraction (MVC) signal of the wrist flexors was determined for normalization of the EMG signals. Muscles were considered

active when the linear envelope exceeded 10 percent MVC.

A split plot analysis of variance (ANOVA) was used to detect group and velocity effects and group and slope effects ($p<0.05$). Differences between peak angles and peak angles with wrist flexor activity were tested with a paired T-test ($p<0.05$).

RESULTS

Global inspection of the data revealed an individually consistent movement pattern of the wrist (**Figure 2**) during most of which the wrist flexors were active. The push phase was started in combined radial deviation-extension and changed into ulnar deviation-flexion. Peak wrist angles occurred at or around hand contact (hc) and hand release (hr). There was, however, a considerable interindividual variation (**Figure 3**).

As can be seen in **Table 2**, no differences were found between the peak wrist angles for the WU and control subjects, apart from a significant difference between the peak flexion values. Palmar flexion was $-3\pm 14^\circ$ for the control group and $-28\pm 8^\circ$ for the WU

Table 3.

Peak wrist angles, averaged over N=9 subjects and six experimental conditions. Peak values are calculated over the full push phase and over that part of the push phase during which wrist flexor activity (WFA) was recorded. The numbers in () represent the standard deviation.

	peak angle (°)	peak angle (+WFA) ^a (°)	Difference (°)
flexion ($\alpha_{w,1}$)	-14 (18) (N=54)	-16 (15) (N=43)	1.5 ^{ns} (N=43)
extension ($\alpha_{w,1}$)	34 (16) (N=54)	32 (16) (N=43)	1* (N=43)
ulnar deviation ($\gamma_{w,2}$)	-24 (11) (N=54)	-22 (11) (N=48)	3** (N=48)
radial deviation ($\gamma_{w,2}$)	13 (12) (N=54)	13 (10) (N=48)	2** (N=48)

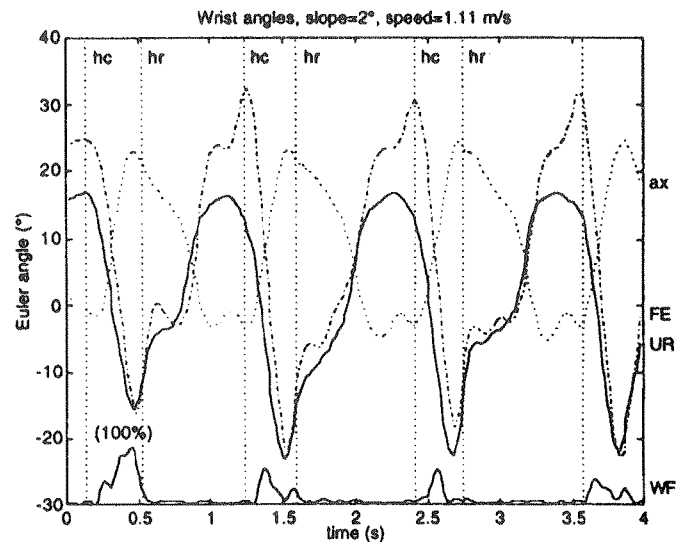
ns, *, **, indicate level of significance (p<0.05, p<0.01)

group. Mean peak values for deviation ranged from $-23 \pm 9^\circ$ to $15 \pm 12^\circ$ for the controls and $-24 \pm 12^\circ$ to $12 \pm 11^\circ$ for the WUs. Wrist flexion-extension ranged from $-3 \pm 14^\circ$ to $36 \pm 19^\circ$ and $-28 \pm 8^\circ$ to $30 \pm 12^\circ$ for the control and WU groups respectively. Extension was significantly influenced by treadmill speed and showed a highest value at 4 km/hr. In addition, small but significant speed and slope effects were found for ulnar deviation.

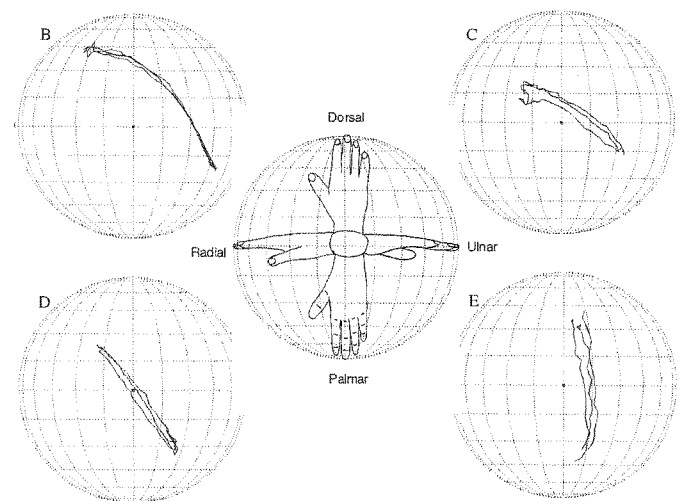
WFA was assumed to be present when EMG values were higher than 10 percent of their MVC value. Wrist motions in all directions occurred simultaneously with WFA.

Comparison of EMG data with wrist motion data, showed that for most subjects and conditions, wrist motions in all directions occurred simultaneously with WFA. In absolute terms, peak extension during WFA was the largest ($32 \pm 16^\circ$, **Table 3**). Although the differences between absolute peak angles and peak angles with WFA were small (maximum 3°), all differences were significant, except for peak angles during WFA (**Table 3**). The finding that the wrist flexors were mostly active, implies that even during extreme wrist motions other than flexion, the wrist flexors were active.

As can be concluded from the mean values and standard deviations in **Table 2**, several subjects reached ulnar deviation angles exceeding the reported mean

**Figure 2.**

Typical example of the wrist angles during three consecutive wheelchair pushing cycles. hc=instant of hand contact, hr=instant of hand release. FE, UR, and ax are the flexion-extension, ulnar-radial deviation, and axial rotations, respectively. The bottom graph illustrates wrist flexion (normalized and rectified). The 100 percent MVC level is indicated in the graph.

**Figure 3.**

Trajectory of wrist motion during the push phase. Plotted is the projection of MCIII on a sphere centered around the middle of the wrist joint. A (center) illustrates the movement directions, and B, C, and D show three consecutive cycles from three different WU subjects; E is from a control subject. *=neutral position; the outer circle represents 90° .

active maximum of -35° reported in the literature (**Table 4**) in one or more conditions. One subject

exceeded -35° in four out of six conditions. The same applied for radial deviation where the criterion value of 21° was also regularly exceeded. None of the subjects reached the criterion value for flexion. Only one subject exceeded the criterion value for extension (66°) in some conditions. As a result, the average peak values, reported in **Table 3**, are close to the reported ROM for ulnar and radial deviation. The average peak values for extension were approximately 50 percent of the ROM. Palmar flexion is relatively close to neutral.

DISCUSSION

In handrim wheelchair propulsion several factors can be distinguished that lead to CTS:

- extreme wrist excursions during the push phase
- finger flexor activity necessary for gripping the rim
- direct pressure of the rim to the carpal tunnel
- highly repetitive character of the movement.

In this study, we recorded the wrist movement pattern of nine subjects during handrim wheelchair propulsion. Five were controls and four were WUs. In previous research, differences were found between controls and WUs in the gross movement pattern during handrim wheelchair propulsion (19,20). Comparisons between controls and WUs concerning wrist joint angles have not been made before.

Wrist angles were defined as Euler angles of the order $x-z'-y''$ (Equation 4), where the first rotation around the x -axis represents flexion-extension and the second rotation around the z' -axis represents ulnar-radial deviation. The rotation around the y'' -axis represents a rest angle, around the longitudinal axis of the hand.

Although the wrist can possibly be described as a joint with two degrees of freedom (DOF; 21), a decomposition in Euler angles will result in three rotations, as long as the local coordinate system is not aligned along the 'anatomical' axes. The magnitude of the third rotation is thus an indication of the accuracy with which the local coordinate system is aligned along the 'true' axes, and of course the accuracy of the movement analysis. In this study the range of the tertiary angle was $17 \pm 11^\circ$, which can be considered as reasonably small. Better alignment of the local coordi-

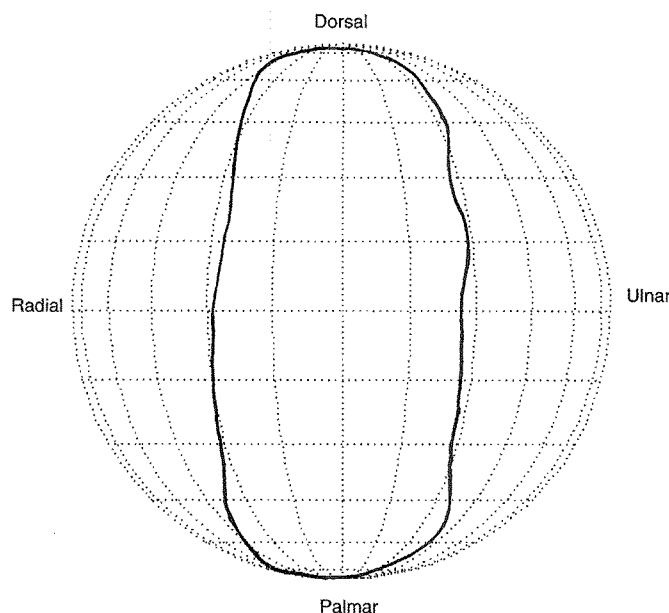


Figure 4.

Simultaneous maximal wrist angles as determined by Braune and Fischer (1887). Adapted from Fick (1911). Projection of MCIII on a sphere centered around the middle of the wrist joint. The center is the middle position.

nate system to the anatomical axes might reduce this range.

The decomposition of wrist angles in a primary flexion-extension angle and a secondary ulnar-radial deviation angle led to minor differences in the deviation values, when compared to primary deviation angles.

Despite the fact that wheelchair propulsion is a guided movement that prevents large variations in technique, considerable differences in wrist movements and in peak angles existed between subjects (**Figure 3**, **Table 2**). In this study, a large difference in peak flexion was found between controls and WUs (-3° versus -28°). The results for the control group were comparable with the flexion-extension angles reported by Boninger et al. (11), who studied a group of six WU subjects, whereas the peak values for wrist flexion reported by Rao et al. (12) were in the midrange (17.9° , $N=16$ WUs). The range within the control group ($\sim 40^\circ$) is also comparable to results reported by Rodgers et al. (10) and Rao et al. (12) for WU populations, who reported ranges of approximately 40° and 49° , respec-

Table 4.
Maximal active range of wrist motion.

	Boone and Azen ⁽¹⁴⁾	Ryu et al. ⁽¹⁵⁾	Sinelnikoff and Grigorowitsch ⁽¹⁶⁾	AAOS ⁽¹⁷⁾	mean
ulnar deviation (°)	-35	-38	-38	-33	-35
radial deviation (°)	21	21	24	19	21
flexion (°)	-75	-79	-78	-73	-76
extension (°)	74	59	63	71	66

tively. Radial-ulnar deviation values did not differ between the control and WU groups and were in a range of approximately 41°. The range was comparable with results by Boninger et al. (11), but smaller than reported by Rodgers et al. (10): approximately 70° and larger than reported by Rao et al. (12): 30.3°.

Interpretation of differences between studies is, however, difficult, given the fact that data reported by Rodgers were based on a 2-DOF model of the wrist joint in which pro-supination of the forearm was neglected, and a smaller diameter handrim (38 cm) used. Rao et al. did not report the handrim diameter, but used a 3-DOF model for the wrist joint, as was done in this study, but did not specify the anatomical position. This might have led to a different 'neutral' position in their subjects, when compared to this study.

The finding that there were some differences between controls and WUs suggests that a careful consideration is necessary when choosing subjects for an experiment. However, comparison of our data with those by Boninger et al. (11) seems to imply that differences between control and WU subjects are not simply related to difference in physical status or experience between those groups, but might be due to the large variations within control and especially WU groups (12). The general conclusion that control subjects should not be used in biomechanical analyses of wheelchair propulsion (19) therefore seems to be overly simplified.

The difference in flexion found between the control and WU groups could have been the result of a difference in gripping style. Different gripping styles will require different degrees of alignment of the hand to the rim and will, therefore, result in different wrist angles at similar push angles. An enclosing grip or power grip (22) will force a full alignment of the hand to the rim throughout the whole push phase, whereas a contact grip (22) will require a good alignment only at

the first part of the push phase. This difference will result in a wide variation of wrist angles at the same push angle in the last part of the push. It is possible that differences between subjects were caused by these different strategies.

Only few studies (14–17) describe the active ROM for wrist angles. On average, reported flexion-extension values range from -76° to 66° (**Table 4**). For ulnar deviation/radial deviation, those values range from -35° to 21°. Usually, these ROM values have been reported as rotations around one axis, without interaction, whereas in wheelchair propulsion, wrist deviation and flexion take place simultaneously. Braune and Fischer (23, in 24) published simultaneous maximal wrist angles (**Figure 4**). From this figure, it can be concluded that peak wrist deviation varies only slightly by alteration of the flexion angle. Therefore, comparison of the multiple wheelchair propulsion values with nonsimultaneous maximal joint angles appears to result in minor errors.

Wrist angles often rose above the ROM values reported in the literature. This especially was the case with ulnar and radial deviation (**Tables 3 and 4**). Of course, one of the reasons that wrist angles often rise above the literature values for maximal ROM might be related to the guided character of the movement, which could force the wrist into excursions larger than those that would have been measured as maximal active excursions.

In absolute terms, peak extension during WFA was the largest (31°, **Table 3**) but peak radial and ulnar deviation during WFA were considerably closer to the maximum ROM values. All peak angles during WFA were within 3° of the absolute peak values. This implies that, in general, even during extreme wrist motions the wrist flexors were active.

Repeated ulnar deviation has regularly been named as an important risk factor for CTS (12,23). From the

study by Phalen (25) that reported a much higher pressure in the carpal tunnel in extension than in flexion, it can be concluded that wrist extension is also an important risk factor. Armstrong et al. (7) suggest that large wrist angles, combined with finger flexor activity, are even more harmful than large angles by themselves. The stretched tendons of the muscles will curve around the carpal bones, resulting in a radial force that could cause micro-damage to tendons and tendon sheaths (7). Due to the shape of the carpal tunnel, the radius of tendon curvature is smaller in deviation when compared to flexion (26). At a given muscle force, this results in a tangential force at least equal to that in wrist flexion, with risk for damage even at smaller deviations. Therefore, the large peak extension during WFA, as well as the smaller peak radial and ulnar deviations during WFA (**Table 4**), could in time be detrimental to the wrist and lead to CTS. Ergonomic recommendations for hand tools generally advise to hold the wrist in a neutral position and suggest allowable ranges of 15° for flexion and extension, 5° for radial deviation, and 10° for ulnar deviation (27).

Recently, Boninger et al. (11) estimated the peak net reaction acting force at the wrist to be approximately 30 N. Although potentially harmful, these reaction forces can be considered as quite low when compared to forces that can be expected during transfers, when the reaction forces are, by definition, at least 50 percent of body weight, that is, 10 times as high. The actual magnitude of the compression in the carpal tunnel is, however, dependent on both the reaction force, the type of grip, and the wrist position. Armstrong and Chaffin (28) stated that the use of a power grip would be a lower risk factor for CTS than a contact grip. In the latter technique, direct pressure on the region of the carpal tunnel will be necessary to create sufficient friction between hands and rim.

To reduce the harmfulness of wheelchair propulsion several suggestions have already been made. Among these are other propulsion systems like lever, crank, or hubcrank (29) and the use of gloves (3). It is difficult to suggest a change in handrim grip. An enclosing grip versus a contact grip will most likely reduce direct pressure on the carpal tunnel, whereas the range of wrist angles is likely to be larger. The contact grip might lessen the forced character of the movement, giving the opportunity of positioning the wrist with less ulnar deviation. This might, however, lead to larger direct pressure on the carpal tunnel. The effectiveness of the choice for a type of grip and for the related decrease

of peak joint angles, direct external force, and WFA should be evaluated as well as their possible effects on efficiency, power production, and other aspects of manual wheelchair propulsion.

CONCLUSIONS

During wheelchair propulsion, near maximal deviations as well as large flexion-extension angles simultaneous with wrist flexor activity are found. These angles exceed values that are generally used as risk limits. It is concluded that the movement pattern is a serious risk factor for CTS. This finding might explain the high prevalence rates of CTS in the WU population (3,4).

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